2007 IEEE International Conference on
Robotics and Automation
Roma, Italy, 10-14 April 2007

Contact Probe Based Stiffness Sensing of Human Eye by Considering Contact Area

Yuichi Kurita, Yoshichika Iida, Roland Kempf, Makoto Kaneko,
Hidetoshi Tsukamoto, Eiichiro Sugimoto, Seiki Katakura, and Hiromu K Mishima

Abstract— **The contact tonometer is commonly used for measuring the internal eye pressure to diagnose glaucoma. However, the conventional eye pressure measurement is valid only under the assumption that all subjects have the same structural eye stiffness. This paper challenges to measure the contact area between the probe and the cornea in addition to the corneal deformation for considering the individual differences in the structural eye stiffness. Prior to the experiment, a spherical model of an eye is developed and the analytical eye stiffness is introduced. The experiment is conducted based on the contact method where a contact probe is pressed on an anesthetized cornea. The deformation of the cornea and the contact area are captured by cameras during the experiment. The experimental results show that the measured stiffness nicely matches the analytical solution based on the constructed model. However, some subjects have different relationships between the displacement and the contact area even with similar estimated eye pressures. This suggests that the structural eye stiffness should be considered for more precise diagnoses of glaucoma.**

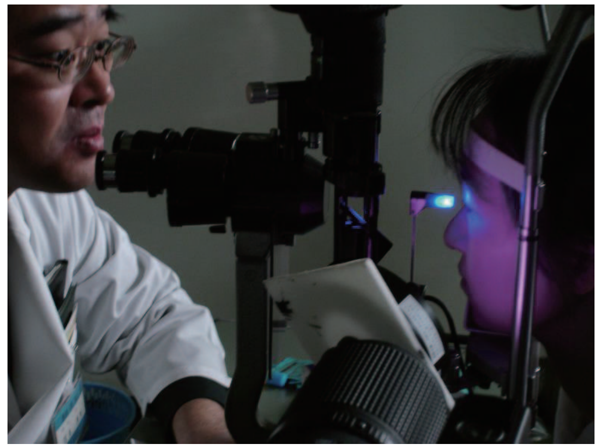


Fig. 1. Contact probe based eye pressure measurement

I. Introduction

For the diagnosis and the treatment of glaucoma, the internal eye pressure is an important parameter. Instead of the invasive method, where a micro needle is inserted into the eye, non-invasive methods are commonly used for measuring the internal eye pressure. In contact tonometry [1], the internal eye pressure is estimated by observing the deformation of the cornea when an external force is applied. The conventional method is valid for the majority of patients, but fails for some patients, who might have non-standard structural eye stiffness such as corneal thickness and elasticity. In order to guarantee the estimation accuracy of the eye pressure, it is necessary to obtain more details on the structural eye stiffness.

The total stiffness of a human eye including the structural stiffness can be measured if both the applied force and the displacement are known. Kaneko et al. [2], [3] have captured the dynamic deformation of the cornea and evaluated the eye stiffness based on the non-contact method, which uses an air jet to deform the cornea. They assumed that the effects coming from both the viscosity and the inertia of the eye deformation are sufficiently small in comparison with the effect of the stiffness. However, the measured eye stiffness changes with time and it indicates that the non-contact based eye stiffness includes time-dependent effects affected by the viscosity and the mass. Kurita et al. [4] have also measured the eye stiffness by using the contact method, which uses a contact probe to deform the cornea as shown in Fig. 1, for eliminating the time-dependent effects. They have revealed that the relationship between the applied force and the displacement is linear under the pressure application in steps of 0.005[N] from 0.005 to 0.03[N].

The purpose of this paper is measuring more precise eye stiffness based on the contact method and evaluating the individual difference of the eye stiffness. At first, a simple spherical eye model is proposed for calculating the analytical eye stiffness when a static force is applied on an eye with an internal pressure. Next, the deformation measurement system for human eyes composed of a contact-type tonometer (eye pressure estimation equipment) and a high resolution camera is presented. The applied pressure is increased gradually and continuously from 0.005 to 0.025[N]. The contact area between the probe and the cornea is simultaneously measured. The contact eye stiffness is defined based on the applied force and the bending of the cornea. The experimental results show that the measured stiffness nicely matches the analytical

This work was supported by JSPS the 21st Century COE program: "Hyper Human Technology toward the 21st Century Industrial Revolution" and MEXT KAKENHI Grant-in-Aid for Young Scientists (B) 18700412.

Y. Kurita, and Y. Iida are with the Graduate School of Engineering, Hiroshima University, 1–4–1 Kagamiyama, Higashi–Hiroshima, Hiroshima 739–8527, Japan kurita@hfl.hiroshima-u.ac.jp

R. Kempf is with Institut für Automatisierungstechnik, TU Darmstadt, Landgraf-Georg-Str.4, D–64283, Darmstadt, Germany

M. Kaneko is with the Graduate School of Engineering, Osaka University, 2–1 Yamadaoka, Suita, Osaka 565–0871, Japan

H. Tsukamoto, and S. Katakura are with the Graduate School of Biomedical Sciences, 1–2–3 Kasumi, Minami–Ku, Hiroshima 734–0037, Japan

E. Sugimoto is with Chugoku Rousai Hospital, 1–5–1 Hirotagaya, Kure, Hiroshima 737-0193, Japan

H. K Mishima is with Hiroshima General Hospital of West Japan Railway Company, 3–1–36 Futabanosato, Higashi–Ku, Hiroshima 732–0057, Japan

solution.

II. Related Works

A contact type tonometer (eye pressure estimation equipment) [1] provides us with a good example of static based sensing, where a medical doctor presses a rigid probe to an anesthetized eye. The non-contact type tonometer [5] is perhaps the closest one to dynamic sensing, while the estimation method is still based on a static strategy. Although the contact type tonometer provides us with relatively accurate values due to the direct contact, the non-contact type tonometer is commonly used in the medical institutions because anesthesia of patients' corneas is not needed and automatic measurement systems are commercially available.

Unfortunately, it is known that the estimated eye pressure depends on the method used [6]. Moreover, individual differences of corneal properties also affect the estimation of the internal eye pressure. Since both the contact and the non-contact method are based on the assumption that all the patients have the same structural corneal properties, the individual differences cause an error in the eye pressure estimation. Some researchers have investigated the effect of the differences in the corneal properties on the eye pressure estimation [7], [8]. For example, it is well known that the corneal thickness affects the estimation of the eye pressure [9], [10]. The same applies to the elasticity of the cornea, which was measurement in vitro [11], [12]. Consequently, several papers propose statistic based correction methods of the estimated eye pressure considering the corneal properties [13], [14].

There are few works that investigate the stiffness of human eyes in vivo. In recent years, Pallikaris et al. [15] have measured the corneal rigidity. Kaneko et al. [2], [3] have measured the total eye stiffness in vivo by using a non-contact method and Kempf et al. [16] have shown that the principle shape of the deformation can be understood by assuming simple non-linear material properties.

An accurate model of a human eye has also been constructed for more precise diagnosis of eye diseases by using FEM models [17], [18]. However, the result has not been compared with the experimental result of a real human eye.

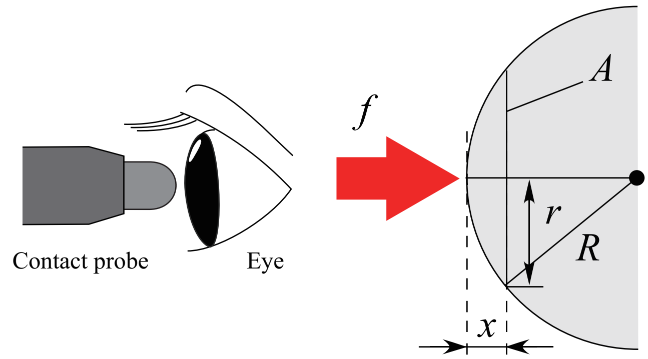
III. Analytical model and eye stiffness

A simple spherical eye model that has internal pressure p is considered, as shown in Fig. 2 where R, f, x, A, and r are the curvature radius of the cornea, the applied force, the bending of the cornea, the deformed area due to the applied force, and the radius of the deformed area. Let us assume that the eye does not change its internal pressure due to the deformation and the cornea has neither bending stiffness nor flexibility.

From Fig. 2 (b), we can formulate the following geometrical relationship:

$$R^2 = (R - x)^2 + r^2. \tag{1}$$

Since x is typically smaller than 0.3[mm] and R is in the range of $7 \sim 8$[mm], we can assume that x is sufficiently



(a) Actual (b) Spherical model

Fig. 2. Model of a human eye

smaller than R and derive the following relationship among r, R, and x:

$$r^2 \simeq 2Rx. \tag{2}$$

Multiplying both sides by πp and using $\pi r^2 = A$ results in

$$f = pA = 2\pi Rpx. \tag{3}$$

From eq. (3), the eye stiffness k is derived by the following equation:

$$k = \frac{f}{x} = 2\pi Rp. \tag{4}$$

Eq.(4) indicates that the eye stiffness k is proportional to the internal pressure p.

For example, the eye with a corneal curvature of $R = 7.78$[mm] and an internal pressure of $p = 14.5$[mmHg], which are the average for all the subjects in our measurements, may have the following eye stiffness:

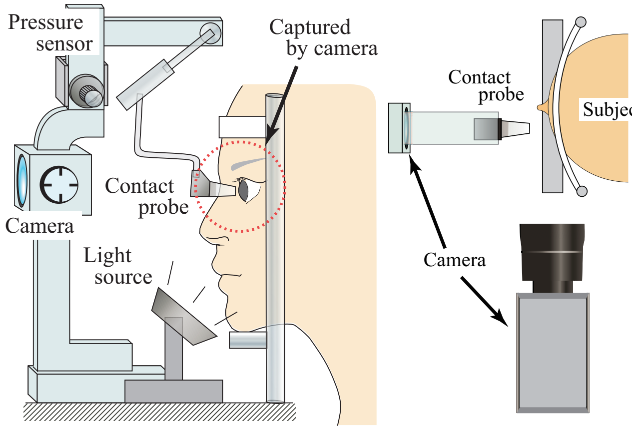
$$k = 2\pi Rp = 2\pi \times 7.78 \times 10^{-3} \times 14.5 \times 133 \simeq 94.3\text{[N/m]}. \tag{5}$$

This discussion provides us with a good hint for evaluating the validity of experimental results.

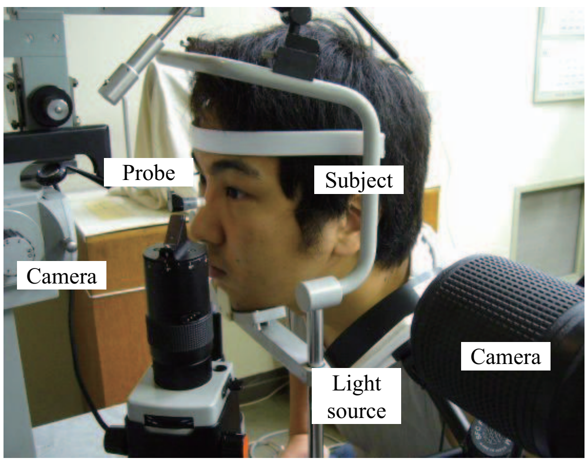
IV. Experimental System

A. System Configuration

The constructed experimental system is composed of a contact-type tonometer and a high resolution camera. The contact-type tonometer uses a contact probe to deform the anesthetized cornea. The applied force to the cornea is controlled by rotating a pressure dial attached on the tonometer. The deformation is captured by the high resolution camera (Flovel co., Ltd.: ADP-210B). The high resolution camera has a spatial resolution of 5.6[μm/pixel] and an image size of 1624×1234[pixels]. The contact area between the transparent probe and the cornea is captured by another camera attached on the tonometer. The camera has a spatial resolution of 19[μm/pixel] and an image size of 640×480[pixels]. The images of the deformation and the contact area are simultaneously stored in a PC with a time resolution of 5[Hz].



(a) Side view (b) Top view



(c) Overview

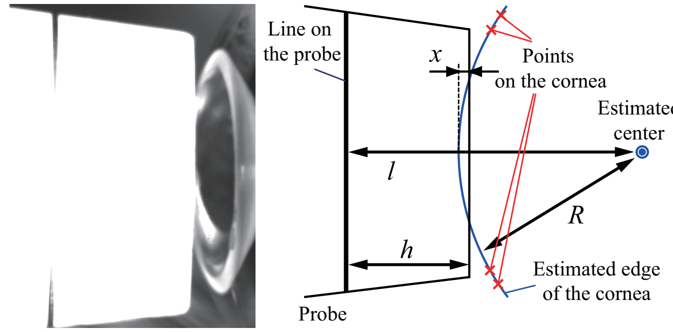
Fig. 3. Experimental system

B. Measurement of the displacement

Fig. 4(a) shows a captured image of the deformed cornea by the contact probe. Because the whole eye ball moves backward while being pressed by the probe, the displacement detection method should consider the change in the position of the eye ball in order to extract the displacement due to the bending of the cornea. In this study, the edge of the corneal surface is detected by an image processing method for eliminating the effect of the whole eye motion.

We assume that the corneal surface is spherical. Based on some points on the fringe of the cornea obtained from the captured images and the curvature radius of the cornea, which has been measured by a medical equipment (Carl Zeiss Co., Ltd.: IOLMaster) prior to the experiment, the edge of the corneal surface and the center of the corneal curvature can be estimated. The overlapping area of the estimated edge and the contact probe is the bending area of the cornea. We define the tip displacement of the bending area as “displacement”.

Fig. 4 (b) shows the diagram of Fig. 4 (a) where h, R, l, and x are the distance between the line on the contact probe and the head of the probe, the curvature radius of the cornea, the distance between the line on the probe and the center of the corneal curvature, and the tip displacement, respectively.



(a) Captured image (b) Diagram

Fig. 4. Detection of the displacement

Since the position of the line on the probe can be clearly observed in the images, l can be computed based on the estimated center of the corneal curvature. h is a known geometrical parameter. Thus the tip displacement x can be computed based on the following geometrical relationship:

$$x = (h + R) - l. \tag{6}$$

V. Experiment

A. Subjects and procedure

35 subjects aged 19∼48 years old (20 males and 15 females) participated in the experiment. The experimental procedure is almost the same as in a normal contact tonometry. A medical doctor pressed the contact probe to the subject’s cornea, which was anesthetized by an eye lotion. The applied force was measured by the attached encoder on the force controller, and the deformation of the cornea and the contact area between the probe and the cornea were captured by the cameras. The applied force was gradually increased from 0.005[N] to 0.025[N]. The curvature radius of the cornea of each subject was measured by IOLMaster prior to the experiment.

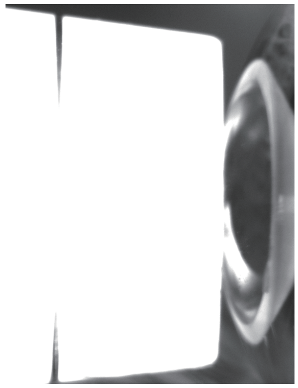
B. Experimental results

Fig. 5 shows representative pictures of the cornea during the experiment: (a) before the contact, (b) when a force of 0.005[N] is applied, (c) when a force of 0.015[N] is applied, and (d) when a force of 0.025[N] is applied, respectively. The displacement of the cornea tip is computed based on the described method in the section IV-B. Six representative examples of the relationship between the applied force and the tip displacement are shown in Fig. 6. We can observe a linear relationship between the applied force and the displacement. This well agrees with the analytical relationship between f and x shown in eq. (3).

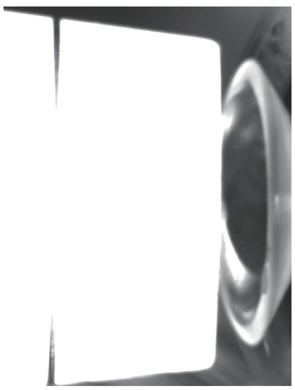
The overlapped line in Fig. 6 (f) shows the approximation line detected by the least square method. In this study, the contact eye stiffness k_{cnt} is defined by the increases of the applied force and the displacement:

$$k_{cnt} = \Delta f / \Delta x \tag{7}$$

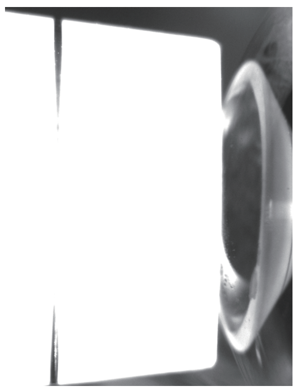
(a) Before the contact



(b) $f = 0.005$[N]

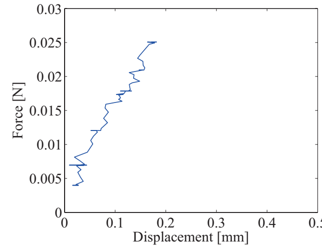


(c) $f = 0.015$[N]

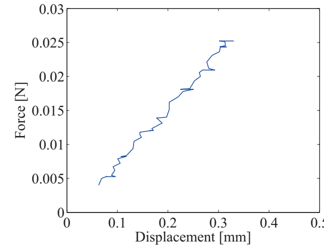


(d) $f = 0.025$[N]

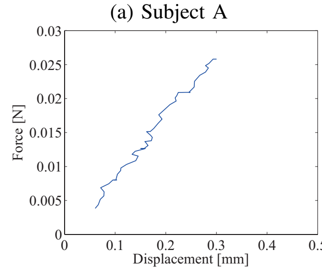
Fig. 5. Deformation of the eye by the probe contact



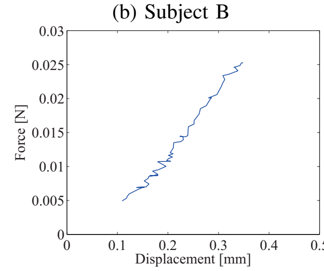
(a) Subject A



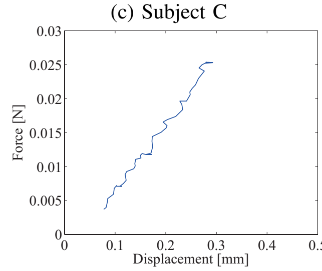
(b) Subject B



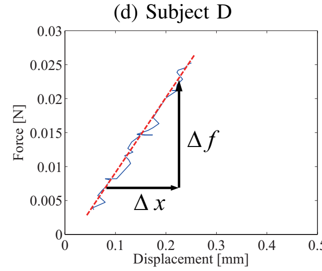
(c) Subject C



(d) Subject D



(e) Subject E



(f) Subject F

Fig. 6. Displacement of the cornea tip due to the force application

where Δf is the increase of the applied force and Δx is the increase of the displacement. Fig. 7 shows the relationship between the estimated eye pressure by the contact tonometer and the computed contact eye stiffness for all the subjects. A good correlation can be observed between them (the correlation coefficient is 0.633, $p < 0.0001$).

Fig. 8 shows the captured contact area between the probe and the cornea. The centers of the upper and the lower circles in the images are shifted by 3.06[mm]. Medical doctors actually use these contact images for detecting the eye pressure. The computed contact area based on these images is shown in Fig. 9 for representative two subjects whose estimated eye pressures are 14[mmHg] and 18[mmHg]. The relationship between the applied force and the contact area is also linear and it well agrees with the analytical relationship of $f = pA$.

VI. Discussion

The analytical solution of the eye stiffness described in the section IV-B depends on the curvature radius of the cornea. We normalize the stiffness by the curvature radius in order to compare the analytical values with the experimental values. The following relationship can be derived from eq. (4):

$$k/R = 2\pi p. \tag{8}$$

Fig. 10 shows the relationship between the normalized stiffness k_{cnt}/R and the pressure related value $2\pi p$ based on the measured contact stiffness k_{cnt}, the curvature radius of the cornea R, and the estimated eye pressure p.

The dashed line in the figure indicates the analytical solution. We can observe the analytical relationship nicely matches the experimental result. The correlation coefficient between the pressure related value and the normalized stiffness is 0.621 ($p < 0.0001$).

Our deformation model used for the calculation of the analytical eye stiffness ignores some important factors. The real cornea has thickness and it contributes to the bending stiffness and the elasticity of the cornea. The tear film causes the tension between the contact probe and the cornea. Moreover, when an eye ball is deformed, the tissue in the eye ball absorbs the aqueous fluid and the internal eye pressure itself might be changed. In addition to the image noises, these factors seem to affect the computation of the eye stiffness value. However, the measured values of the eye stiffness by the contact method ($60 \sim 140$[N/m]) is closer to the analytical results than the stiffness values measured by the non-contact method ($500 \sim 3500$[N/m]) [3].

Here we discuss the difference of the stiffness value between the contact method and the non-contact method. The main difference of these methods is how the force is applied to the cornea. The deformation starts, when the external force generated a sufficient pressure on the area

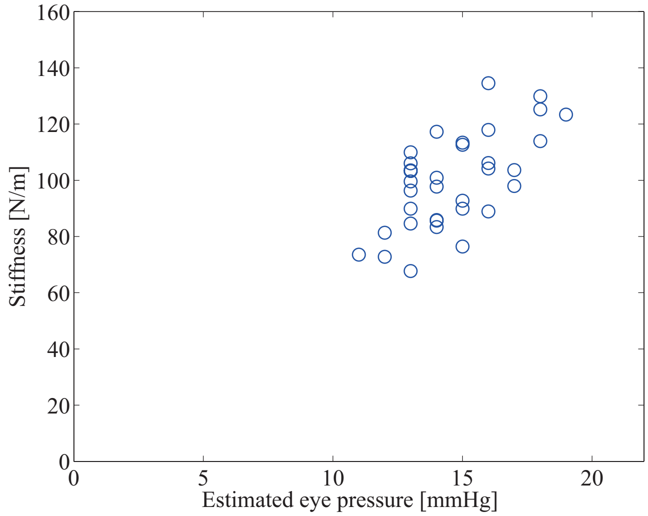


Fig. 7. Correlation between the calculated contact stiffness and the estimated eye pressure

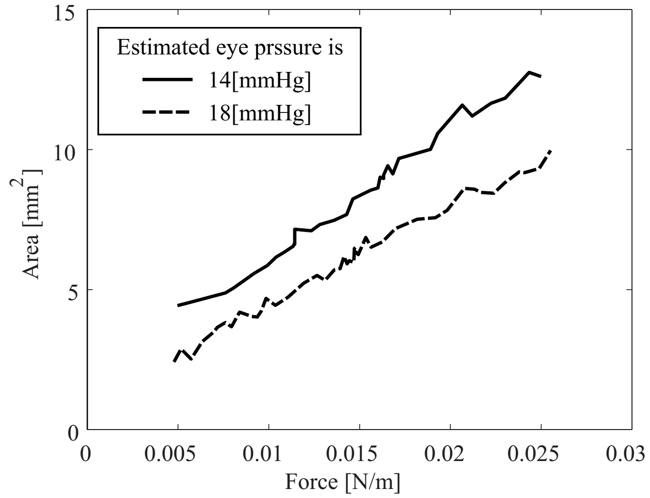
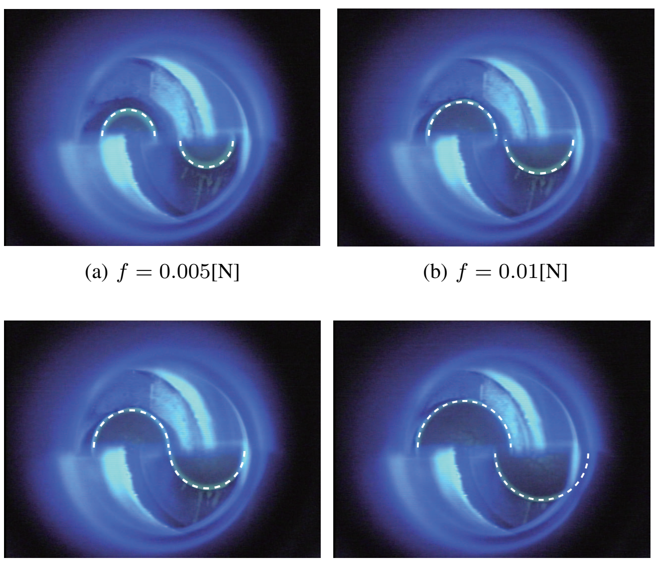


Fig. 9. Relaionship between the applied force and the contact area for two subjects.



(a) $f = 0.005$[N]

(b) $f = 0.01$[N]

(c) $f = 0.015$[N]

(d) $f = 0.025$[N]

Fig. 8. Contact area between the probe and the cornea whose estimated eye pressure is 14[mmHg]

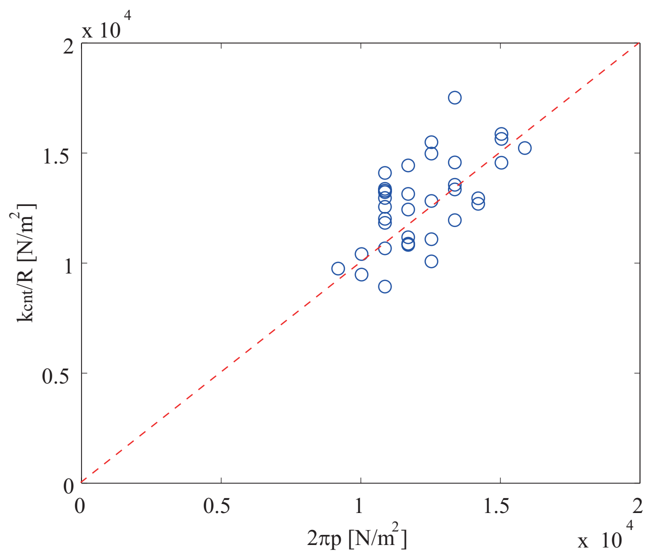


Fig. 10. Normalized stiffness by the curvature radius

of deformation. In contact tonometry, the force is directly applied and the threshold pressure is created in a certain area. In non-contact tonometry, the total force of the air jet is dispersed over the cornea and it generates a pressure distribution on a large area. Thus only a fraction of the applied force is causing a deformation and, as a consequence, the eye stiffness is overestimated. This is one of reasons possible for the difference of the stiffness value.

In addition, note that the contact and the non-contact method have different definitions for the eye stiffness. In the non-contact method, the stiffness is defined based on the force and the displacement at a prescribed time because the applied force dynamically changes depending on the time. In the contact method, the stiffness is defined based on the increase of the applied force and the displacement. Since the stiffness with the different definition has different characteristics, a direct comparison is not possible. The discussion of the difference in the eye stiffness depending on the measurement method will be considered in our future work.

A noteworthy point is that we can observe some subjects have different stiffness with same estimated eye pressure. The subjects with the extraordinarily low or high stiffness might have abnormal structural eye stiffness. It leads to the underestimation or the overestimation of the internal eye pressure. Because the current estimation method only evaluates the applied force and the contact area, the current method can not distinguish the deformations shown in Fig. 11 due to the difference of the structural eye stiffness.

Fig. 12 shows the relationship between the displacement and the contact area for three subjects whose estimated eye pressures are 14, 15, and 18[mmHg] respectively. If the struc-

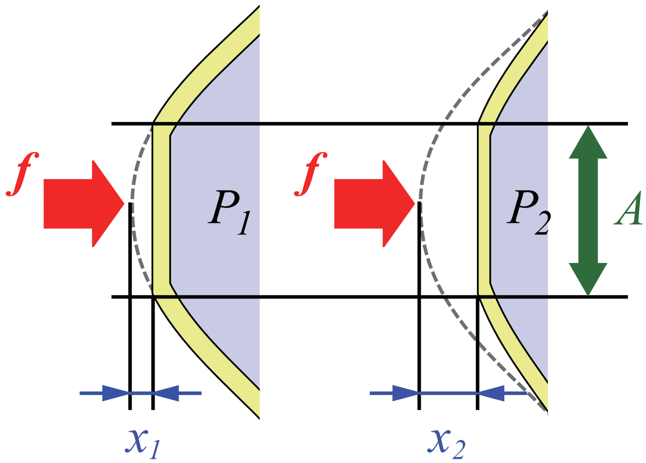


Fig. 11. Expected corneal deformation with same contact area but different displacement. Even if the true internal eye pressure is different ($P_1 \neq P_2$), the current method assumes these two eyes have same internal eye pressure because the deformed area (A) by the same applied force (f) is same.

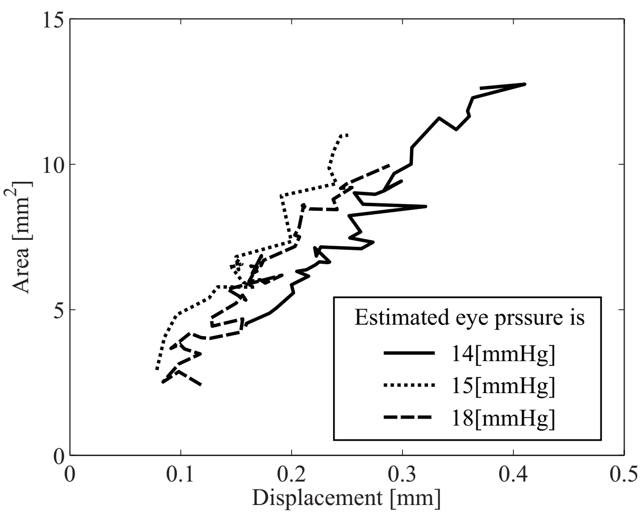


Fig. 12. Relationship between the displacement and the contact area

tural eye stiffness is same for all the subjects, the relationship between the displacement and the contact area should be similar for subjects that have similar internal pressures. In this case, however, the deformation of the eye with the internal pressure of 15[mmHg] is more similar to that with the pressure of 18[mmHg] than that with the pressure of 14[mmHg]. This suggests that the corneal deformation is not uniform and the deformation may be affected by the structural eye stiffness. The evaluation of the eye stiffness in addition to the currently used information might contribute early detection and treatment of glaucoma.

VII. Conclusion

The stiffness information is important for a more precise diagnosis of glaucoma. Although there are some works that report the stiffness information of the eye, these papers have measured the eye stiffness by a dynamic force application. This study is the first work that measures the stiffness of human eyes based on the contact method in vivo.

What we have done in this paper can be summarized as follows:

- The analytical eye stiffness was formulated based on a simple spherical eye model.
- The deformation measurement system of human eyes composed of a contact-type tonometer and a high resolution camera was developed.
- A linear correlation between the applied force, the displacement, and the contact area was observed.
- The experimental values nicely matched the analytical values.

VIII. Acknowledgments

The authors are thankful for Mr. Hiroshi Koizumi (TOPCON Corporation) for his meaningful comments on our experiment.

References

[1] H. Goldmann. Applanation tonometry. In F. W. Newell, editor, *Glaucoma; Transactions of the Second Conference*, pages 167–220. New York: Josiah Macy, Jr, Foundation, 1957.
[2] M. Kaneko, K. Tokuda, and T. Kawahara. Dynamic sensing of human eye. In *Proc. of the IEEE Int. Conf. on Robotics and Automation*, pages 2882–2887, Barcelona, Spain, April 2005.
[3] Y. Kurita, Y. Iida, R. Kempf, M. Kaneko, H. K Mishima, H. Tsukamoto, and E. Sugimoto. Dynamic sensing of human eye using a high speed camera. In *Proc. of the IEEE Int. Conf. on Information Acquisition*, pages 338–343, Hong Kong, China, June 2005.
[4] Y. Kurita, R. Kempf, Y. Iida, M. Kaneko, E. Sugimoto, H. Tsukamoto, and H. K Mishima. Eye stiffness measurement by probe contact method. In *Proc. of the IEEE Engineering In Medicine and Biology Annual Conference*, pages 2312–2315, New York, USA, Sep 2006.
[5] Topcon corporation. *http://www.topcon.co.jp/*.
[6] P-A. Tonnu, T. Ho, K. Sharma, E. White, C. Bunce, and D. Garway-Heath. A comparison of four methods of tonometry: method agreement and interobserver variablity. *Br. J. Ophthalmol.*, 89:847–850, 2005.
[7] Y-C Ko, CJ-I. Liu, and W-M. Hsu. Varying effect of corneal thickness on intraocular pressure measurement with different tonometers. *Eye*, pages 1–6, 2004.
[8] J. Liu and C. J. Roberts. Influence of cornea biomechanical properties on intraocular pressure measurement. *J. Cataract Refract Surg.*, 31:146–155, 2005.
[9] N. Ehlers and F. K. Hansen. Central corneal thickness in low tension glaucoma. *Acta Ophthalmol.*, 54:740–746, 1974.
[10] Y. Morad, E. Sharon, L. Hefetz, and P. Nemet. Corneal thickness and curvature in normal-tension glaucoma. *Am. J. Ophthalmol.*, 125:164–168, 1998.
[11] C. Edmund. Corneal topography and elasticity in normal and keratoconic eyes. *Acta Ophthalmol.*, 44:367–408, 1989.
[12] J. O. Hjortdal. Regional elastic performance of the human cornea. *J. Biomechanics*, 29(7):931–942, 1996.
[13] M. Shimmyo, A. J. Ross, A. Moy, and R. Mostafavi. Intraocular pressure, goldmann applanation tension, corneal thickness and curvature in caucasians, asians, hispanics, and african americans. *Am. J. Ophthalmol.*, 136:603–613, 2003.
[14] G. J. Orssengo and D. C. Pye. Determination of the true intraocular pressure and modulus of elasticity of the human cornea in vivo. *Bull. Math. Biol.*, 61:551–572, 1999.
[15] I. G. Pallikaris, G. D. Kymionis, H. S. Ginis, G. A. Kounis, and M. K. Tsilimbaris. Ocular rigidity in living human eyes. *Investigative Ophthalmology and Visual Science*, 46(2):409–414, 2005.
[16] R. Kempf, Y. Kurita, Y. Iida, M. Kaneko, H. K Mishima, H. Tsukamoto, and E. Sugimoto. In *Proc. of the IEEE Engineering In Medicine and Biology Annual Conference*, pages 5428–5431, New York, USA, Sep 2006.
[17] P. M. Pinsky and D. V. Datye. A microstructurally-based finite element model of the incised human cornea. *J. Biomechanics*, 24:907–922, 1991.
[18] M. R. Bryant and P. J. McDonnell. Constitutive laws for biomechanical modeling of refractive surgery. *J. Biomech. Eng.*, 118:473–481, 1996.